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Co-contraction during static and dynamic knee extensions in ACL deficient subjects

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Abstract

Co-contraction of the muscles is proposed in the literature as one of the strategies that anterior cruciate ligament deficient (ACLD) subjects can use to compensate the loss of ACL function. This study examined the response of ACLD and control subjects to different shear forces in isometric and slow-dynamic knee extensions.

Twelve chronic ACLD and 10 control subjects performed submaximal positioning and slow-dynamic knee extensions (between 45° and 5° of knee flexion) with two external flexion moments both applied at two distances on the lower leg. The shear force was controlled by changing the moment arm without changing the moment. Electromyographic data were collected from knee flexor and extensor muscles.

In the analysis of variance, no significant effect of subject group was found in positioning or slow-dynamic tasks across all muscles. The effect of knee angle was significantly different between the subject groups for biceps femoris in positioning and for rectus femoris in slow-dynamic tasks, but these effects were very small and will not have a great impact on the resulting shear forces. There was no interaction between moment arm and subject group. Therefore, the hypothesis that ACLD subjects increase co-contraction in situations with an increased shear load in positioning and slow-dynamic knee extensions could not be confirmed.

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Keywords: Anterior cruciate ligament; Shear force; Isometric; Slow-dynamic; Co-contraction

1. Introduction

It is well known that after anterior cruciate ligament (ACL) rupture the laxity in the knee is increased and that most patients experience instability of the knee. The ACL is the main ligament to prevent anterior shear displacement of the tibia relative to the femur. ACL deficient (ACLD) subjects lack this restraint against anterior shear displacement. One of the possible strate-

gies that ACLD subjects can use to prevent anterior tibial displacement during knee extension is co-contraction of the Hamstrings.

In recent years, a lot of research has focussed on co-contraction during functional activities [3,5,7,8,11,13,15,20,21,24–27]. In these studies, contradictory results were reported regarding the strategy ACLD and healthy subjects use during functional activities. It is difficult to distinguish co-contraction strategies from other strategies, such as changes in knee extension moment, during these functional activities.

Measuring isokinetic or isometric knee extensions decreases the complexity of the movement and are,

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therefore, useful methods to examine the co-contraction strategy. Only a few isokinetic or isometric studies have been done on ACLD subjects. Osternig et al. [19] found more co-contraction in the injured leg, while others found no co-contraction [17] or no difference in co-contraction between normal and ACLD subjects [9].

The knee can be challenged differently by changing the location of the load on the lower leg, while keeping the required extension moment constant. A more proximal placement of a resistance pad will decrease the tibiofemoral displacement [14] and will lead to less force on the ACL [28]. Kingma et al. [16] used this method to examine co-contraction in various shear challenges for the ACL in healthy subjects. They found no change in co-contraction with a change of the resistance position in isometric knee extension in healthy subjects. However, healthy subjects may not need to prevent force on the ACL in sub-maximal knee extensions. In subject with a ruptured ACL, the need for compensation of the ACL might lead to co-contraction that depends on the shear challenge to the knee joint.

Therefore, the aim of this study is to examine whether ACLD subjects use a different co-contraction strategy compared to healthy subjects in static and dynamic knee extensions with different shear challenges to the knee. The hypothesis is that ACLD subjects will increase co-contraction in situations with an increased shear challenge to the knee joint.

2. Methods

2.1. Subjects

Twelve chronic (>6 months post-injury) anterior cruciate ligament deficient (ACLD) subjects participated in the experiment. In all ACLD subjects, rupture of the ACL was confirmed by MRI scan, arthroscopy or clinical testing. The ACLD subjects were asked to fill out the Lysholm score [23] and the International Knee Documentation Committee (IKDC) subjective questionnaire [12] and scored on average 81.5 (SD 7.8; range 62–92) on the Lysholm and 72.6 (SD 13.6; range 51.7–97.7) on the IKDC. Ten healthy subjects without any knee problems formed a control group. Table 1 shows the subject characteristics for both subject groups. All subjects signed an

informed consent before the measurements. The protocol was approved by the medical ethics review committee of the VU University Medical Center.

2.2. Experimental set-up

The subject was seated in a custom made chair (Fig. 1) with the hip joint in 60° flexion. The knee joint of the leg was aligned with the rotation axis of the loading apparatus. The lower leg was placed in a leg-holder on a rigid arm, which was connected to the wheel. The lower leg mass and the weight of the holder and lever arm were counterbalanced by weights on the opposite side of the wheel. A constant external moment was applied over the range of motion (0–50° of knee flexion). The subject was asked to perform two tasks (positioning and slow-dynamic) in five loading conditions (15 and 30 N m external flexion moment, both applied with a moment arm of 120 and 300 mm, and 25 N m external extension moment applied with a moment arm of 300 mm). A pilot study suggested that a number of untrained ACLD individuals would not be able to maintain 40 N m at 120 mm for 5 s in all knee angles. To avoid dropouts, especially in the ACLD group, the maximal load was set at 30 N m. The external moments were applied through the leg-holder onto the lower leg. The posterior shear force caused by the external force at 120 mm has been calculated to have approximately the same magnitude as the anterior shear force due to quadriceps contraction at flexion between 0° and 45°, so that, even in the absence of hamstrings co-contraction, the extension effort would not cause ACL loading [28]. At 300 mm, the posterior shear force due to the external force is considerably smaller than the forward shear force due to the activation of the quadriceps. This would result in an increased anterior shear force challenge to

Table 1
Subject characteristics

	ACLD	Control	<i>p</i> -Value
Sex (m/f)	7/6	4/6	
Age (year)	29.9 (7.5)	24.1 (4.0)	0.04
Height (m)	1.80 (0.06)	1.79 (0.13)	0.74
Weight (kg)	76.0 (12.1)	72.3 (12.5)	0.39

Values are average (standard deviation).

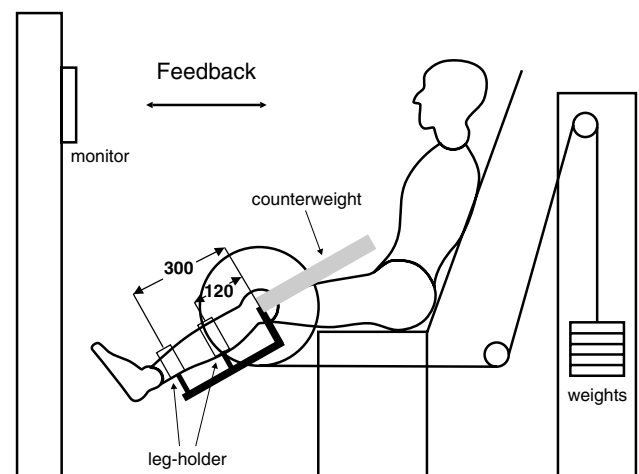


Fig. 1. Schematic overview of the chair. The subject was seated on a chair with the hip in 60° flexion. The subject received real-time feedback on the knee joint angle.

the knee joint (and thus resulting in an increased loading of the ACL in healthy subjects) compared to the external force position close to the joint. In addition, flexion efforts were performed with one moment (-25 N m) and one force position (300 mm). The order of the moments and moment arms was randomised over subjects.

Subjects received real-time position feedback about their knee joint angle. After a practice trial, the subjects were asked to extend the knee once slowly through the full range of motion (slow-dynamic task) and six times to a target angle (positioning task). During the positioning task, the subject was asked to extend the knee to the target angle and hold it there for 5 s. The six target angles were 5° , 10° , 15° , 20° , 30° , and 45° , applied in random order. Subjects were allowed to rest as long as they needed when they felt fatigued.

2.3. Data analysis

An opto-electronic movement recording system (Optotrak, Northern Digital Inc., Canada) was used to give real-time knee angle feedback and display the target angles on a screen in front of the subjects. LED markers were placed on the frontal side of the leg, two on the thigh and two on the tibia. The line indicating zeros degrees of knee flexion on the feedback-screen was aligned with unloaded full knee extension. Marker positions were sampled at 100 Hz and used to calculate angular velocity of the knee extension.

Electromyography (EMG) from six leg muscles (vastus medialis: VM, rectus femoris: RF, vastus lateralis: VL, semimembranosus: SM, semitendinosus: ST, and biceps femoris: BF) was obtained, using disposable surface EMG-electrodes (Ag/AgCl; square $5\text{ mm} \times 5\text{ mm}$ pick-up area). The skin was shaved, abraded and cleaned before electrode attachment and the center-to-center electrode distance was 2.5 cm. Surface EMG locations were based on Seniam guidelines [10]. During the slow-dynamic and the positioning task, EMG signals were sampled at 1000 Hz (Porti 17, TMS, Enschede, The Netherlands; 22 bits AD conversion after $20\times$ amplification, input impedance $>10^{12}\ \Omega$, CMMR $>90\text{ dB}$ for the relevant range of frequencies), and band-pass filtered with 10 and 250 Hz cut-off frequencies. A pulse generated by Optotrak was used to synchronise EMG and Optotrak data in time.

2.4. Data processing

The EMG signals measured during the positioning tasks were rectified and averaged over 4 s during the isometric part of each trial. In the dynamic tasks, the EMG signals were rectified and plotted against the knee angle after which it was averaged in blocks of 5° between 2.5° and 47.5° . All EMG signals were normalised against the average EMG of all knee angles of the positioning task

at 300 mm moment arm and 30 N m extension moment for extensors and the 300 mm moment arm -25 N m flexion moment for flexors.

2.5. Statistics

For all muscles and for both the positioning and dynamic tasks, repeated measures ANOVA's were applied to the normalised and raw EMG data of the extension efforts. The between-subject variable was subject group (ACLD or healthy) and the within-subject variables were joint angle (six levels for positioning and nine levels for dynamic tasks), moment (two levels), and external force position (two levels). The same ANOVA's were applied to the standard deviation of the joint angle during the recorded period. In cases where significant interactions were found, separate ANOVA's were applied. The smallest detectable difference was calculated for each muscle using a *t* test between groups over all conditions.

3. Results

3.1. Effect of moment and knee joint angle

A statistically significant main effect of moment and knee joint angle was found in all muscles in both the positioning and the dynamic tasks (Figs. 2 and 3). Not surprisingly, an increase in moment led to an increase in Quadriceps EMG, but also to an increase in Hamstrings EMG. A decrease in knee angle led to an increase in EMG in all muscles in both positioning and dynamic tasks. The effect of knee angle was stronger at 30 N m than at 15 N m (Table 2, interaction moment \times knee angle).

3.2. ACLD vs. normal subjects

Raw EMG data showed statistically similar effects as the normalised EMG data. There was no significant main effect of subject group (Table 2; $p > 0.2$) in any of the muscles neither in positioning nor in slow-dynamic tasks. As might be expected, the EMG during slow-dynamic tasks was higher, than during positioning tasks. The angular velocity was quite variable over subjects, and significantly higher in ACLD subjects (ACLD: $14.7^\circ \pm 5.6^\circ$ per second, control: $9.8^\circ \pm 3.6^\circ$ per second; Table 2; $p = 0.005$). However, both the main effects of moment and moment arm and the interactions with subject group were not significant. The positioning data (Fig. 2) suggests a weak tendency in ACLD subjects to apply more VM, VL and BF co-contraction compared to healthy subjects in higher knee angles, and less co-contraction in lower knee angles. The interaction between subject group and knee angle was only significant

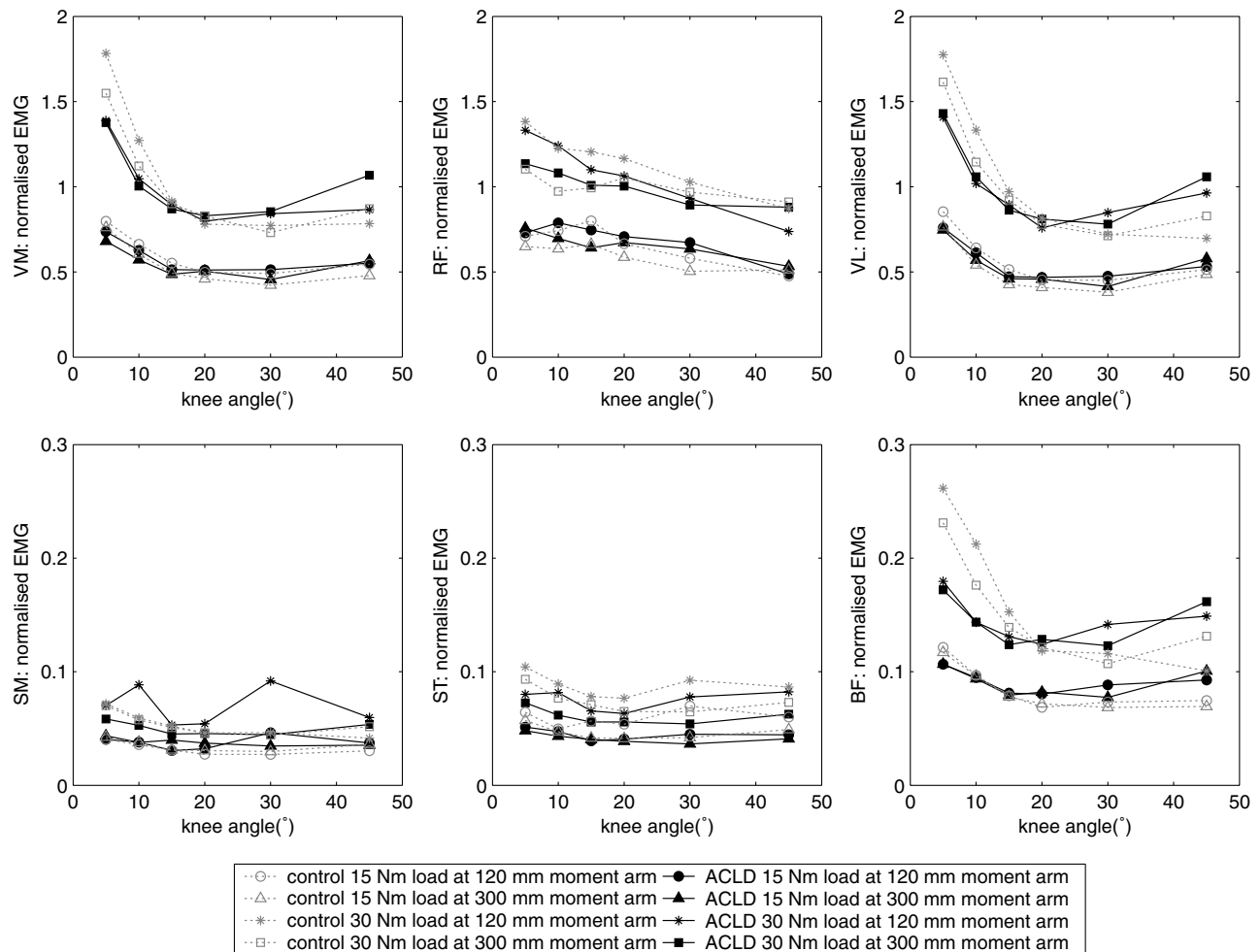


Fig. 2. Positioning tasks: comparing conditions. Normalized EMG of six muscles during positioning tasks for both control and ACLD group. VM, vastus medialis; RF, rectus femoris; VL, vastus lateralis; SM, semimembranosus; ST, semitendinosus; BF, biceps femoris.

for BF in positioning and for RF in slow-dynamic tasks (Table 2, Fig. 3). In both cases, separate tests per knee angle showed no significant effect of subject group in this muscle. The BF also showed an interaction between moment, knee angle and subject group for the positioning tasks (Table 2). Individual results at 300 mm moment arm (averaged over moments) show large inter-individual differences (Fig. 4). The smallest detectable difference between the ACLD group and the control group ranged from 3.0% to 5.0% of 25 N m for the Hamstrings and 5.9% to 7.4% of 30 N m for the Quadriceps muscles.

3.3. Response to shear force challenge

The shear force challenge was controlled by changing the moment arm, without changing the external moment. A main effect of moment arm was only found in the RF muscle in the positioning tasks and in the BF muscle in dynamic tasks. Interestingly, neither moment arm nor moment arm \times knee angle interacted with sub-

ject group, indicating a comparable strategy between the control group and the ACLD group in response to different shear force challenges to the knee joint. In addition, moment arm modified the effect of knee angle (interaction effect moment arm \times knee angle) for the VM, RF, VL and BF muscles in both the positioning and the dynamic tasks. Again, it should be noted that there was no interaction with subject group, suggesting a comparable strategy between the ACLD and the control group. The tests per knee angle showed a significant effect of moment arm in the positioning tasks in the following knee angles: 5° (VM, RF, BF), 10° (VM, RF, BF), 15° (RF), 20° (RF) and 30° (VL, RF). In the dynamic tasks, the effect of moment arm was mainly seen in larger knee angles: 30° (VL), 35° (VL, BF), 40° (VM, VL, BF) and 45° (VM, VL, SM, BF).

The EMG during slow-dynamic tasks showed an interaction effect of moment \times moment arm in 5 out of 6 muscles (Table 2, Fig. 3). There was a significant effect of moment arm at 30 N m load for VM ($p = 0.042$), VL ($p = 0.019$) and BF ($p = 0.011$), but none at 15 N m load.

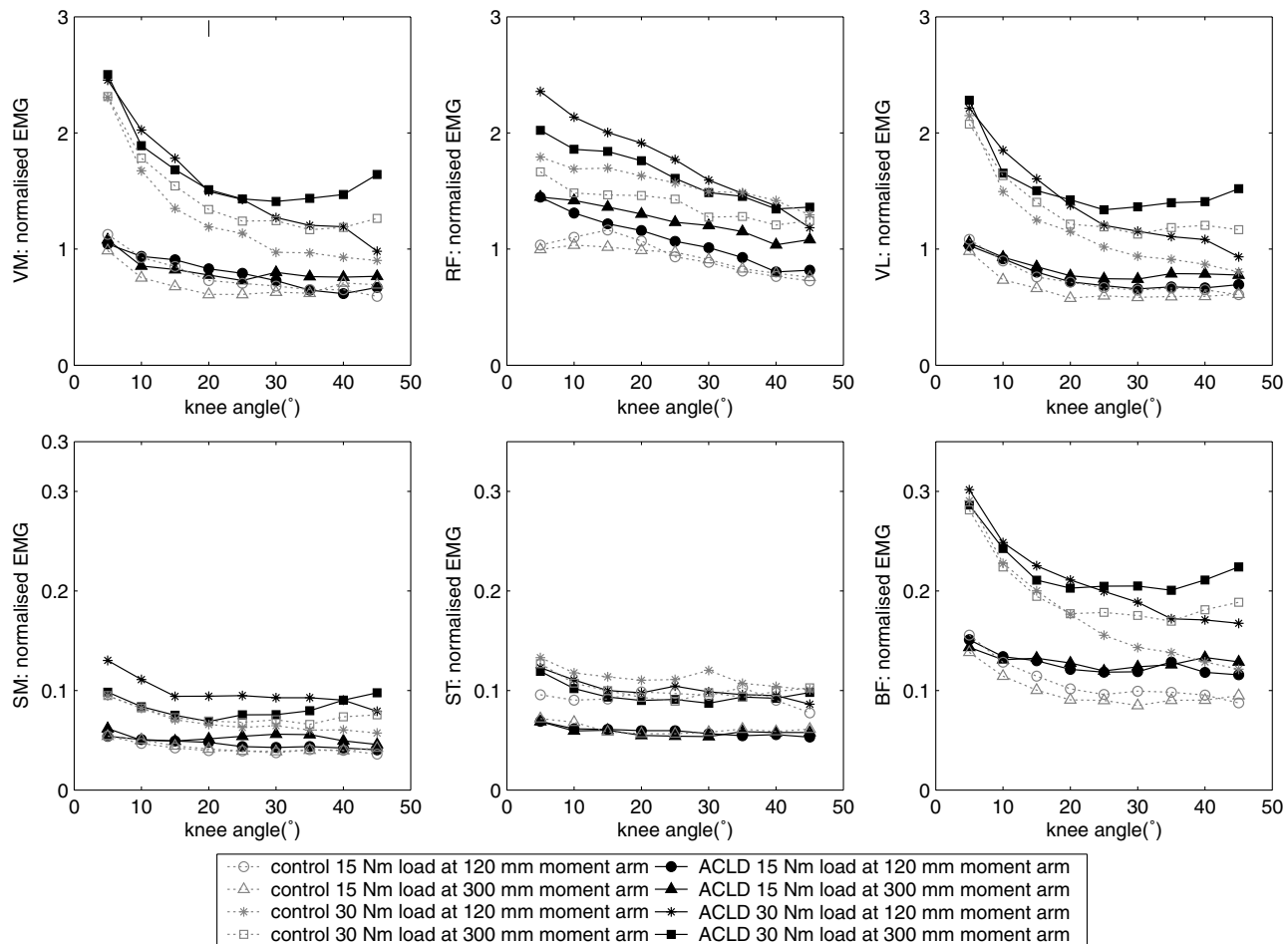


Fig. 3. Dynamic tasks: comparing conditions. Normalized EMG of six muscles during dynamic tasks for both control and ACLD group. VM, vastus medialis; RF, rectus femoris; VL, vastus lateralis; SM, semimembranosus; ST, semitendinosus; BF, biceps femoris.

4. Discussion

An increase in co-contraction was seen near full extension in all loading conditions and both subject groups, which is in accordance with previous findings in healthy subjects in isometric [16] and slow-dynamic [1] knee extensions. This study extends this finding to ACLD subjects. The patella tendon angle is largest at 0° and decreases linearly between 0° and 45° [2,4]. Therefore, the anterior shear load increases with knee extension. Several studies have predicted that hamstring muscles cannot produce effective shear forces in knee angles smaller than 15° [6] and 22° [18]. The reason is that the increase in flexion moment caused by the Hamstrings activation needs to be compensated by the Quadriceps and this will increase the amount of anterior shear force. This increase will be larger than the posterior shear force produced by the Hamstrings. Therefore, the increase in co-contraction in smaller knee angles cannot be seen as a method to directly counteract the anterior shear force produced by the Quadriceps. Increasing co-contraction in knee angles of 30° and

45° seems to be an effective method to counteract anterior shear forces. In the current study, a tendency towards a decrease in co-contraction in smaller knee angles and towards an increase in co-contraction in larger knee angles was found in the ACLD group compared to control group. This can be seen as an adjustment of the co-contraction, although differences between ACLD and control were small. The largest difference in the BF muscle between the ACLD and control group, roughly 5% of the 25 N m flexion effort, is seen at 120 mm moment arm and represents approximately 1.3 N m. Therefore the adjustment is unlikely to have a major impact on the resulting joint shear force.

Loads placed more distally on the lower leg will lead to more anterior shear force in the knee compared to proximally placed loads. In healthy subjects, this shear challenge can be counteracted by the ACL. After an ACL rupture, the shear force challenge cannot be counteracted by the ACL. However, ACLD subjects do not use a strategy that differs from normal subjects in response to variations in shear force challenge during positioning and slow-dynamic knee extensions. The

Table 2
p-Values of the ANOVA tests

Repeated measures ANOVA	VM		RF		VL		SM		ST		BF		α SD		ω	
	pos	dyna	pos	dyna	pos	dyna	pos	dyna	pos	dyna	pos	dyna	pos	dyna	pos	dyna
<i>Between-subjects variables</i>																
Main effect Group	0.854	0.245	0.785	0.364	0.876	0.201	0.859	0.826	0.328	0.382	0.917	0.658	0.800	0.005		
<i>Within-subjects variables</i>																
Mom	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	0.005	0.941		
Mom \times Group	0.487	0.352	0.204	0.724	0.451	0.424	0.969	0.829	0.518	0.510	0.436	0.920	0.389	0.769		
Mom arm	0.223	0.151	0.007	0.620	0.180	0.078	0.490	0.196	0.072	0.215	0.124	0.046	0.005	0.995		
Mom arm \times Group	0.644	0.729	0.388	0.370	0.336	0.405	0.291	0.699	0.439	0.148	0.824	0.396	0.143	0.857		
Knee angle	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	0.013	0.041	0.010	0.011	< 0.001	< 0.001	0.220	–		
Knee angle \times Group	0.101	0.712	0.454	0.008	0.173	0.753	0.272	0.466	0.587	0.678	0.021	0.358	0.821	–		
Mom \times Mom arm	0.696	0.018	0.080	0.049	0.603	0.013	0.243	0.916	0.282	0.040	0.161	0.011	0.006	0.856		
Mom \times Mom arm \times Group	0.770	0.612	0.701	0.636	0.952	0.502	0.350	0.444	0.377	0.139	0.990	0.507	0.811	0.435		
Mom \times Knee angle	< 0.001	< 0.001	0.032	0.003	< 0.001	< 0.001	0.183	0.489	0.032	0.003	< 0.001	< 0.001	0.701	–		
Mom \times Knee angle \times Group	0.103	0.727	0.352	0.074	0.124	0.936	0.298	0.635	0.404	0.149	0.003	0.700	0.217	–		
Mom arm \times Knee angle	0.007	0.001	< 0.001	0.034	0.016	0.004	0.285	0.047	0.192	0.056	0.016	< 0.001	0.701	–		
Mom arm \times Knee angle \times Group	0.722	0.256	0.958	0.477	0.145	0.660	0.373	0.702	0.561	0.415	0.224	0.941	0.586	–		
Mom \times Mom arm \times Knee angle	0.002	0.366	0.043	0.607	0.462	0.078	0.291	0.156	0.501	0.291	0.236	0.007	0.692	–		
Mom \times Mom arm \times Knee angle \times Group	0.129	0.242	0.975	0.250	0.076	0.145	0.624	0.667	0.699	0.567	0.325	0.781	0.554	–		

Mom, moment; Mom arm, moment arm; pos, positioning tasks; dyna, slow-dynamic tasks; α SD, standard deviation of knee angle; ω , angular velocity; bold print, significant effect; VM, vastus medialis; RF, rectus femoris; VL, vastus lateralis; SM, semimembranosus; ST, semitendinosus; BF, biceps femoris.

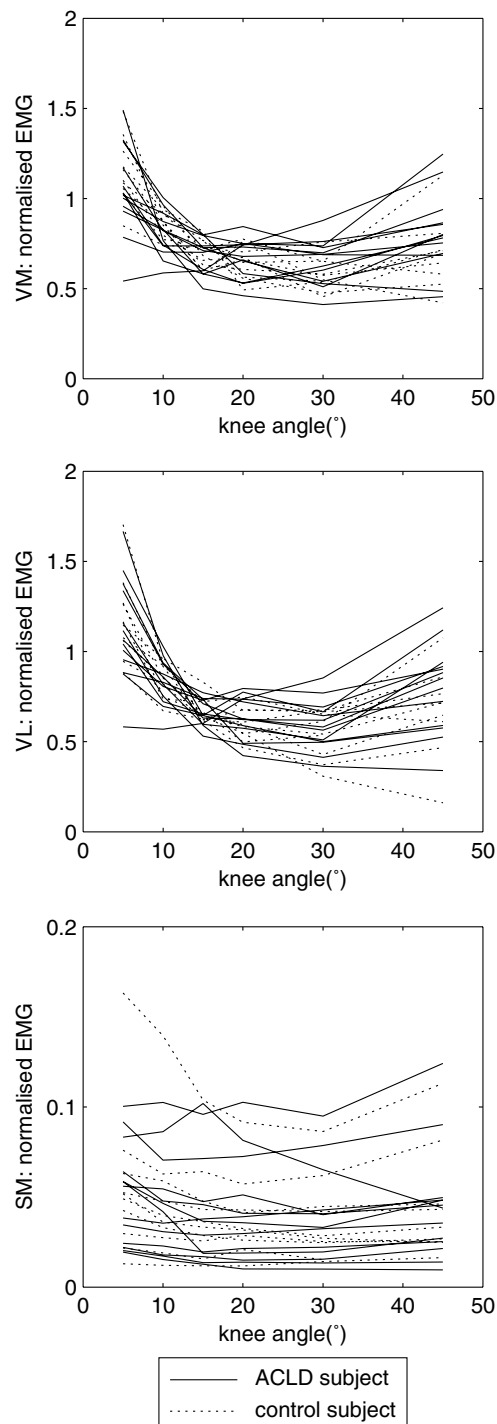


Fig. 4. Positioning tasks: three examples of EMG data of all individual subjects. VM, vastus medialis; VL, vastus lateralis; SM, semimembranosus.

hypothesis that ACLD subjects increase co-contraction in situations with an increased shear challenge to the knee joint could not be confirmed in this study. One of the reasons might be that ACLD subjects allow shear displacement in the knee. Kvist et al. [17] have examined anterior shear displacement during isokinetic knee

extensions and they found that the absolute translation was larger in ACLD subjects. By allowing shear displacement, secondary restraining structures can balance the shear force. Another mechanism may be the decrease in anterior shear force by changing the angle of the patella tendon. Shelburne et al. [22] calculated that anterior tibial translation in the knee model without an ACL decreased the amount of anterior shear force produced by the patella tendon.

The SM and ST EMG ranged from about 2.7% to 10.4% of the EMG seen in 25 N m flexion trials and the BF EMG ranged from 6.8% to 26.2% of the EMG seen in 25 N m flexion trials. The BF EMG was averaged over all trials $2.15 (\pm 0.34)$ times the averaged EMG of the SM and ST. A stronger EMG in the lateral (BF) compared to medial (SM and ST) hamstrings EMG was also found in isokinetic knee extensions [1] and isometric knee extensions [16]. Aagaard et al. [1] suggested that the reason for the more pronounced lateral compared to medial co-contraction might be prevention of internal rotation of the tibia, thereby decreasing the strain on the ACL. An increased interest in preventing internal rotation after an ACL rupture can be expected, but no difference in the ratio of medial and lateral hamstrings EMG was found between the control and the ACLD group. There was no shift towards more lateral hamstrings activity and therefore no increased protection against internal rotation in ACLD subjects. Both the main ACL-injury mechanisms and the pivot shift that occurs after an ACL rupture have an internal axial rotational component. In the current study, the axial rotation of the lower leg was free and no axial rotational loads were applied. A protective strategy against rotation forces might have been more pronounced when a rotational load would have been applied.

Because of the constant loading (as opposed to constant velocity in isokinetic tasks), the angular velocity of the dynamic tasks could not be applied externally and was therefore not constant between conditions and subject groups. The significant differences in angular velocity between subject groups make a direct comparison of EMG levels between subject groups difficult in the slow-dynamic tasks, but despite these variations, the EMG curves did not differ significantly between subject groups. The angular velocity did not differ between moments and moment arms within one subject group. Therefore, comparing the different conditions is justified.

A limitation of this study is that the movements used in this study involved only isometric and slow-dynamic contractions at relatively low contraction levels. During highly dynamic and powerful conditions co-contraction might still be a strategy used by ACLD subjects to prevent shear displacement. Another limitation is that EMG data were not normalised to maximal voluntary contractions (MVC). An MVC is affected greatly by

pain (or fear of pain) and if subjects are not contracting maximally during the MVC measurement, the normalised EMG data will be an overestimation of the actual signal. By using the averaged EMG signals of one moment and moment arm combination, the amount of net moment was known and constant over all subjects. Since both raw and normalised EMG data showed similar statistical effects, the normalisation method used in the current study had no influence on the statistical outcome.

In conclusion, ACLD subjects seem to do approximately the same as control subjects when a shear challenge is applied to the knee. Although the effect of joint angle on co-contraction differed between ACLD and control subjects in one muscle, this difference was small in terms of moment and, therefore, also in terms of potential counteraction of shear forces. Moreover, a direct manipulation of the shear forces resulted in only small changes in co-contraction, and these changes did not differ between ACLD and control subjects. Therefore, within the range of knee joint angles used in this study, there are little to no indications of a different co-contraction strategy between ACLD and control subjects to counteract shear forces during submaximal isometric and slow-dynamic tasks.

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